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Determination of the vertical ground reaction forces acting upon individual limbs during healthy and clinical gait

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ABSTRACT

In gait lab, the quantification of the ground reaction forces (GRFs) acting upon individual limbs is required for dynamic analysis. However, using a single force plate, only the resultant GRF acting on both limbs is available.

The aims of this study are (a) to develop an algorithm allowing a reliable detection of the front foot contact (FC) and the back foot off (FO) time events when walking on a single plate, (b) to reconstruct the vertical GRFs acting upon each limb during the double contact phase (DC) and (c) to evaluate this reconstruction on healthy and clinical gait trials.

For the purpose of the study, 811 force measurements during DC were analyzed based on walking trials from 27 healthy subjects and 88 patients. FC and FO are reliably detected using a novel method based on the distance covered by the centre of pressure. The algorithm for the force reconstruction is a revised version of the approach of Davis and Cavanagh [24]. In order to assess the robustness of the algorithm, we compare the resulting GRFs with the real forces measured with individual force plates. The median of the relative error on force reconstruction is 1.8% for the healthy gait and 2.5% for the clinical gait. The reconstructed and the real GRFs during DC are strongly correlated for both healthy and clinical gait data ($R^2 = 0.998$ and 0.991, respectively).

1. Introduction

Human walking is characterized by the occurrence of the double contact phase (DC), when both feet are on the ground, separating periods of single contact when the contralateral limb is swung forward. The evaluation of the external forces acting upon each lower limb may be required, for example to estimate the joint forces and moments developed at the ankle, knee and hip by the inverse dynamic method [1]. This is classically used for the evaluation of healthy adults [2], advanced age [3], or patients [4]. From a practical perspective, the decomposition of the ground reaction forces (GRFs) into left and right profiles acting upon each limb during DC can be challenging since the subject must perform two consecutive steps with feet on separate force plates. Generally, this implies the subject to target the force plates using visual guidance to place the feet correctly. Even if the variability of GRFs is not affected by 'targeting' [5], kinematic adaptations were shown previously; the motion of the body segments are flawed [6] and the variability of the step length is increased [7]. A counteract is to perform as many trials as necessary but a high number of trials could cause fatigue and could result also in a gait pattern alteration [8]. Moreover, the 'targeting' or the repetition is laborious for the evaluation of patient suffering from neurological or orthopedic disorders.

To overcome this methodological weakness, one solution is to record separately the left and right GRFs by means of a split-belt force treadmill [9,10]. However, this belt design usually constrains the subject to walk with an unnaturally wider base of support [11,12]. Another solution is to use a single belt force treadmill [11,12] or multiple force plates that would measure the sum of GRFs (F_{sum}) and use an algorithm to reconstruct, during DC, the left and right force profiles. However, to get individual GRFs, it is required to first, detect both events defining the beginning and the end of DC (the front foot contact (FC) when the heel strikes the ground and the back foot off (FO) when the toes take off the ground), and secondly, reconstruct the GRFs acting upon each limb by means of an efficient algorithm.

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Additional equipment is often used to detect FC and FO gait events using cameras [13–15], sensors on the sole or the insole of the shoe [16,17], miniature gyroscopes [18] or accelerometers [19]. The precision of the detection is highly variable with these additional measurements and they present at least two disadvantages: a potential discomfort for the subject, and synchronization between devices can be problematic.

More interestingly, the FC and FO gait events can also be detected using only the force plates measurement and without any additional material. For instance, various authors [20–22] determined FC and/or FO by locating a minimum or an inflexion point on either F_{sum} traces or its derivative. Although these detections seem convenient, this inflection point is not systematically present in GRF recordings and they are partly based on expected durations or expected individual GRF patterns. As already spotted by Ballaz et al. [23], those methods are not usable in many cases and particularly when the walking pattern is abnormal and cannot be anticipated. Alternatively, several authors chose to detect FC and FO events using the centre of pressure (CoP). Indeed, in any walking trial, the CoP is relatively constant during single contact phases while it shows large variations during DC, which makes the CoP a very efficient parameter for the FC and FO detection. Davis and Cavanagh [24] determined FC and FO by a visual observation of the CoP while Ballaz et al. [23] and Villeger et al. [25] went one step further using a threshold value. However, the DC detection can be improved, intending to provide an automated and reliable tool without threshold value.

Finally, different methods to reconstruct the individual GRFs during DC have been suggested in the literature. These methods mainly focus on the largest GRF component, the vertical, even though the fore–aft and lateral components may also be needed in analyses like the inverse dynamic. For instance, Davis and Cavanagh [24] proposed a method based on the lateral CoP location and GRFs by means of two simultaneous equations; Begg and Rahman [20] used the forces measured during the previous step, Ballaz et al. [23] used a cubic spline to approximate the vertical GRFs acting upon the back limb during DC, and Villeger et al. [25] enhanced the original algorithm of vertical GRF reconstruction developed by Ren et al. [26] by including pre-DC GRF characteristics and walking speed. However, none of these methods were evaluated on a large sample of healthy or pathological subjects or at different walking speeds. Additionally, although the pioneering reconstruction method of Davis and Cavanagh is the most attractive, some adaptations of their algorithm are needed. Indeed, when a subject walks with a narrowed sustentation base, the feet tend to be aligned, the left foot lever arm becomes equivalent to the right foot lever arm and their equations can no longer be solved.

The aims of this study are the following; (a) to develop an algorithm allowing an automatic, systematic and reliable detection of FC and FO, (b) to improve the Davis and Cavanagh's vertical GRF reconstruction algorithm and (c) to evaluate this new procedure on a large number of steps collected in healthy subjects and patients.

2. Methods

2.1. Subjects

Twenty-seven young healthy subjects (mean age: 22.8 ± 2.6 years, weight: 71.6 ± 9.5 kg) and 88 patients (33 adults, mean age: 40.2 ± 15.5 years, weight: 67.2 ± 17.7 kg and 55 children, mean age: 9.5 ± 3.0 years, weight 31.2 ± 13.4 kg) were enrolled in the study. The inclusion criteria for the healthy subjects were; age over 18 years, no current locomotor system injury complaints, and no history of neurological disorder. Before the experiments, the purpose and the nature of the study were explained to the subjects. Patients' data are issued from the gait laboratory of Saint-Luc university Hospital (Brussels, Belgium) and collected between 2000 and 2002 for medical evaluations. All experiments were performed according to the Declaration of Helsinki and were approved by the local ethics committee.

Table 1
Diagnoses of the patients.

	Numbers of patients	Numbers of steps
Neurological diseases		
Periventricular leukomalacia	2	10
Idiopathic toe-walker	2	11
Hemiplegia	24	117
Diplegia	7	42
Cerebral palsy	29	126
Arnold-Chiari malformation	1	5
Myopathy	1	7
Paraparesis	3	17
Childhood polio	1	5
Quadriparesis	2	12
Orthopedic diseases		
Ankle sprain	2	10
Osteoarthritis of the knee	1	4
Lower limb fracture	1	7
Spondylolysis	1	7
Equinovarus	4	16
Foot valgus	1	11
Spasticity	1	8
Others diseases		
Fibromyalgia	2	10
Walking unstable	2	9
Psychomotor retardation	1	3
Total	88	437

The subjects were asked to walk across a force platform. The mean walking speed was measured by two photocells placed at the beginning and the end of the walkway. A total of 374 steps were recorded for healthy gait where subjects were requested to walk at different speeds ranging from 0.83 to 1.94 m s^{-1}. A total of 437 steps were recorded for clinical gait (352 steps for neurological disorders; 63 steps for orthopedic disorders; and 22 steps for other disorders, Table 1). In all cases, the patients were recorded at their spontaneous walking speed.

2.2. Force platform measurement

The GRFs were measured by means of a force platform mounted at floor level and embedded in the center of a long walkway. All GRFs are reported according to the ISB referential recommendation [27]. The individual signals of the force plates data were collected independently with three different set-ups: either eight 1000 mm × 1000 mm force plates as described in Genin et al. [28], four Arsalis 800 mm × 500 mm force plates as described in Pozo et al. [29] or seven force plates of different sizes as described in Detrembleur et al. [30]. The amplified signals of the force plates were processed by means of a computer with dedicated software. For each of these set-ups, the sample rate was, respectively, 500, 250 and 100 Hz. A complete step cycle was selected for analysis only when the feet were on different force plates in order to compare the real and the reconstructed individual GRFs.

2.3. Calculation of the real FC, FO and GRFs with two force plates

The real vertical component of the GRF acting upon the back limb (F_{back}) and the front limb (F_{front}) were measured from two individual force plates. The real FC timestamp was determined when F_{front} exceeds 10 N (two times the maximum of noise signal) and the real FO timestamp when F_{back} falls below 10 N.

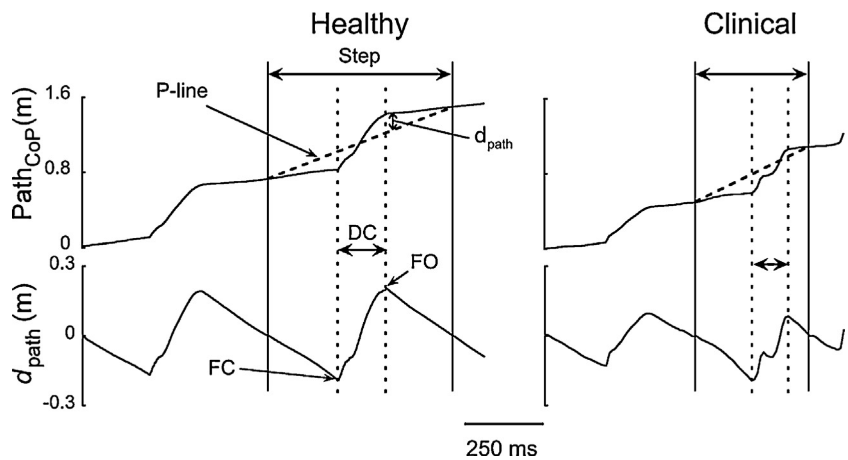

Fig. 1. Detection of FC and FO during walking.
(Top) $Path_{CoP}$: distance covered by the centre of pressure as a function of time. (Bottom) d_{path}: difference between the $Path_{CoP}$ and the P-line as a function of time. Healthy traces are from a 23 year-old subject (weight: 72 kg, height: 1.83 m). The step lasts for 634 ms and is delimited by the continuous vertical lines and the DC lasts for 164 ms and is delimited by the dotted vertical lines. Clinical traces are from a 73 year-old hemiplegic patient (weight: 73 kg, height 1.71 m). The step lasts for 390 ms and the DC lasts for 130 ms.

2.4. Calculation of FC, FO and GRFs with one force plate

In the *one-plate* method, the same GRFs (F_{back} and F_{front}) are summed as if they were collected from a single force plate. Here, the FC and FO were automatically determined by computation of d_{path}, the difference between $Path_{CoP}$ and P-line (Fig. 1). The $Path_{CoP}$ is the distance covered by the CoP and P-line is the line joining the initial and final $Path_{CoP}$ value during one single step cycle. FC corresponds to the minimum of d_{path} curve and FO to the maximum of d_{path} curve (Fig. 1).

Once the FC and FO events are identified, the vertical GRFs acting upon each limb are reconstructed during DC using an algorithm based on an equivalence of forces and of moments.

(*a*) Equivalence of forces in the vertical direction:

$$F_{sum} = F_{back} + F_{front} \tag{1}$$

where F_{sum}, F_{back}, and F_{front} are the vertical GRFs acting upon both limbs, the back limb and the front limb, respectively.

(*b*) Equivalence of moments around an axis *A* on the force platform top surface horizontal plane, such that (Fig. 2): the axis *A* passes through CoP_{back} and is perpendicular to the line d_{front} that joins CoP_{back} to CoP_{front}. CoP_{back} and CoP_{front} are the application points of F_{back} and F_{front}. Thus, axis *A* is defined with the following equations:

$$m = \frac{CoP_{zback} - CoP_{zfront}}{CoP_{xfront} - CoP_{xback}} \quad \text{and} \quad p = CoP_{xback} - m \times CoP_{zback} \tag{2}$$

where *m* is the slope of *A* and *p* is the intercept of *A* with the *x*-axis. Subscripts *x* and *z* relate to the fore–aft and the lateral position of the CoP, respectively.

The location of CoP_{back} is considered fixed and defined as the mean CoP_{sum} during the 20 ms preceding FC (pre-DC CoP characteristic). Similarly, CoP_{front} is considered fixed and defined as the mean CoP_{sum} during the 20 ms following FO (post-DC CoP characteristic).

- the moment generated around A (M_A) is therefore equal to:

$$M_A = d_{front} \times F_{front} = d_{sum} \times F_{sum} \tag{3}$$

where d_{sum} is the distance between CoP_{sum} and the axis *A*, calculated as:

$$d_{sum} = \frac{m \times CoP_{zsum} - CoP_{xsum} + p}{\sqrt{1 + m^2}} \tag{4}$$

and d_{front} is the distance between CoP_{back} and CoP_{front}.

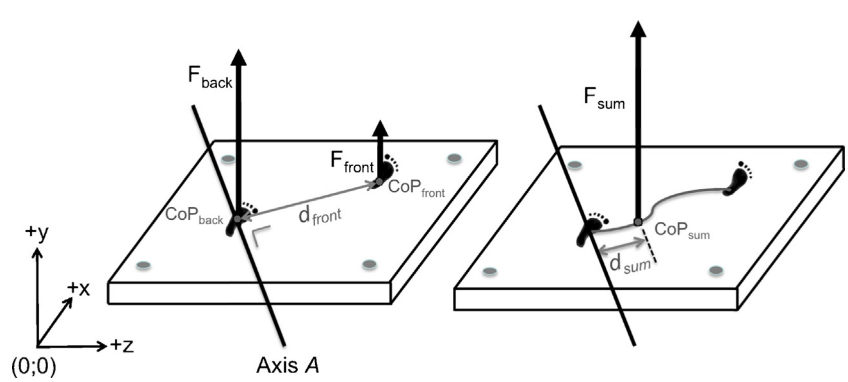

Fig. 2. The equivalence of moments around the axis *A*.
(Left) Illustration of d_{front}. The vertical arrows represent the GRFs acting under the front and back limbs, respectively, F_{front} and F_{back}. (Right) Illustration of d_{sum}. The vertical arrow represents the sum of F_{front} and F_{back} (F_{sum}).
The axis *A* passes through CoP_{back} and is perpendicular to the line d_{front} that joins CoP_{back} to CoP_{front}. d_{sum} is the distance between CoP_{sum} and the axis *A*. GRF referential [27]: +*x* axis is defined as the direction of travel, +*y* axis is defined upward and +z is defined by a right hand rule. The anticlockwise moments are defined as positive.

Finally, by substitution, the vertical GRFs acting under the front and the back limb are obtained by:

$$F_{\text{front}} = \frac{F_{\text{sum}} \times d_{\text{sum}}}{d_{\text{front}}} \quad (5)$$

$$F_{\text{back}} = F_{\text{sum}} - F_{\text{front}} \quad (6)$$

2.5. Calculation of GRF error

The absolute error (e_F), during DC, for GRF acting upon the back limb is calculated as:

$$e_F = \frac{\sum_{i=1}^{n} |F_{\text{reconstructed},i} - F_{\text{real},i}|}{n} \quad (7)$$

where $F_{\text{reconstructed},i}$ is the force calculated at each instant (i) using the reconstruction algorithm, $F_{\text{real},i}$ is the real force and n is the number of samples during DC.

The relative error (ε_F) is calculated as:

$$\epsilon_F = \frac{e_F}{|F_{\max}|} \quad (8)$$

where $F_{\max}$ is the maximal real force during DC and acting under the back limb.

Note that the absolute error calculated for the forces acting on the front limb would be, by mathematics, identical to those calculated for the back limb.

2.6. Comparisons to other methods

In order to compare to our results for FC and FO detections, the methods of Ballaz et al. [23] and of Villeger et al. [25] have been implemented with our data. To compare to our results for the vertical GRF reconstruction, the methods of Ballaz et al. [23] and of Davis and Cavanagh [24] have been implemented with our data.

3. Results

3.1. FC and FO error analysis

The errors of FC and FO detections are calculated by the absolute difference between the real timestamps and the timestamps determined with our method. The distribution of these errors is presented as histograms in Fig. 3A. For the current method, the median value of the error is null for the FC detection and reaches a maximal value of 4 ms for the FO (Table 2).

In order to compare our results to other published methods (Ballaz et al. [23] and Villeger et al. [25]), Table 2 presents the median, the percentile 75 (P_{75}) and the percentile 95 (P_{95}) of FC and FO errors. For FC, all three methods allow a good detection, P_{95} always being less than 20 ms. For FO, our algorithm clearly reduces the error on detection compared to the two other methods. For instance, using the Ballaz et al. [23] and Villeger et al. [25] methods, the FO error is larger than 170 ms (P_{95}) while our method leads to an error of only 40 ms.

3.2. GRF error analysis

A typical trace of vertical GRFs and the relative error (ε_F) as a function of time is shown in the upper part of Fig. 3B. The bottom part of Fig. 3B shows the GRFs obtained by reconstruction as a function of the real GRFs for all DC data samples. The relationship is linear for both healthy and clinical gait data and presents a strong correlation (R^2 equal to 0.998 and 0.991, respectively) with slopes corresponding to the identity line (1.007 and 1.010, respectively).

The median, the P_{75} and the P_{95} of the e_F and ε_F are presented in Table 2 where our results are compared to Ballaz et al. [23] and Davis and Cavanagh [24] methods. The median of ε_F on force reconstruction with the current method is 1.8% for healthy gait and 2.5% for clinical gait. Our algorithm clearly reduces the error on GRF reconstruction compared to the two other methods.

4. Discussion

The goal of this study is to establish and validate a detailed methodology to (a) detect the FC and FO events and (b) decompose F_{sum} into its individual vertical force components (F_{back} and F_{front}).

First of all, a precise time detection of FC and FO is the keystone for an accurate reconstruction of the vertical GRF. As shown in the results, the methods of Ballaz et al. [23] and of Villeger et al. [25] have revealed limits in detecting the DC time events, at least using our data (Table 2). Indeed, those methods failed regularly to identify FO correctly probably because the toe-off transition is less 'noticeable' in the F_{sum} or CoP curves. As a matter of fact, the method of Ballaz et al. [23] is highly dependent on 'ΔCoP^2' thresholds not clearly defined by the authors. More recently,

Table 2
Time events and force reconstruction errors.

Time events error	Healthy gait Median	P_{75}	P_{95}	Clinical gait Median	P_{75}	P_{95}	Total Median	P_{75}	P_{95}
FC (ms)									
Ballaz et al. [23]	12	16	18	0	10	20	10	12	20
Villeger et al. [25]	4	4	6	10	10	10	4	10	10
Current method	**0**	**2**	**4**	**0**	**0**	**10**	**0**	**0**	**10**
FO (ms)									
Ballaz et al. [23]	26	134	190	50	100	170	32	110	184
Villeger et al. [25]	13	34	184	10	30	160	10	30	170
Current method	**4**	**8**	**36**	**0**	**10**	**40**	**4**	**10**	**40**
GRF error									
e_F (N)									
Ballaz et al. [23]	28.3	42.4	69.0	25.3	39.5	82.0	26.7	40.3	73.4
Davis and Cavanagh [24]	57.2	84.1	143.8	18.3	32.6	64.4	33.1	59.1	119.2
Current method	**12.7**	**16.5**	**22.4**	**10.1**	**15.6**	**33.9**	**11.0**	**16.8**	**27.7**
ε_F (%)									
Ballaz et al. [23]	3.8	5.6	9.1	7.0	10.0	17.2	5.3	8.1	14.5
Davis and Cavanagh [24]	7.7	11.1	18.5	4.8	8.3	20.6	6.2	10.0	19.3
Current method	**1.8**	**2.2**	**2.9**	**2.5**	**4.3**	**9.3**	**2.0**	**3.0**	**7.3**

The absolute error on Foot Contact (FC) and Foot Off (FO) detections is expressed in milliseconds. The absolute error during DC on the vertical ground reaction force reconstruction (e_F) is expressed in newtons. The relative error (ε_F) is expressed as a percentage of the maximal GRF acting upon the back limb during DC. Errors are calculated on 374 DC phases for healthy gait and 437 for clinical gait. P_{75} and P_{95} indicate the percentile 75 and percentile 95.

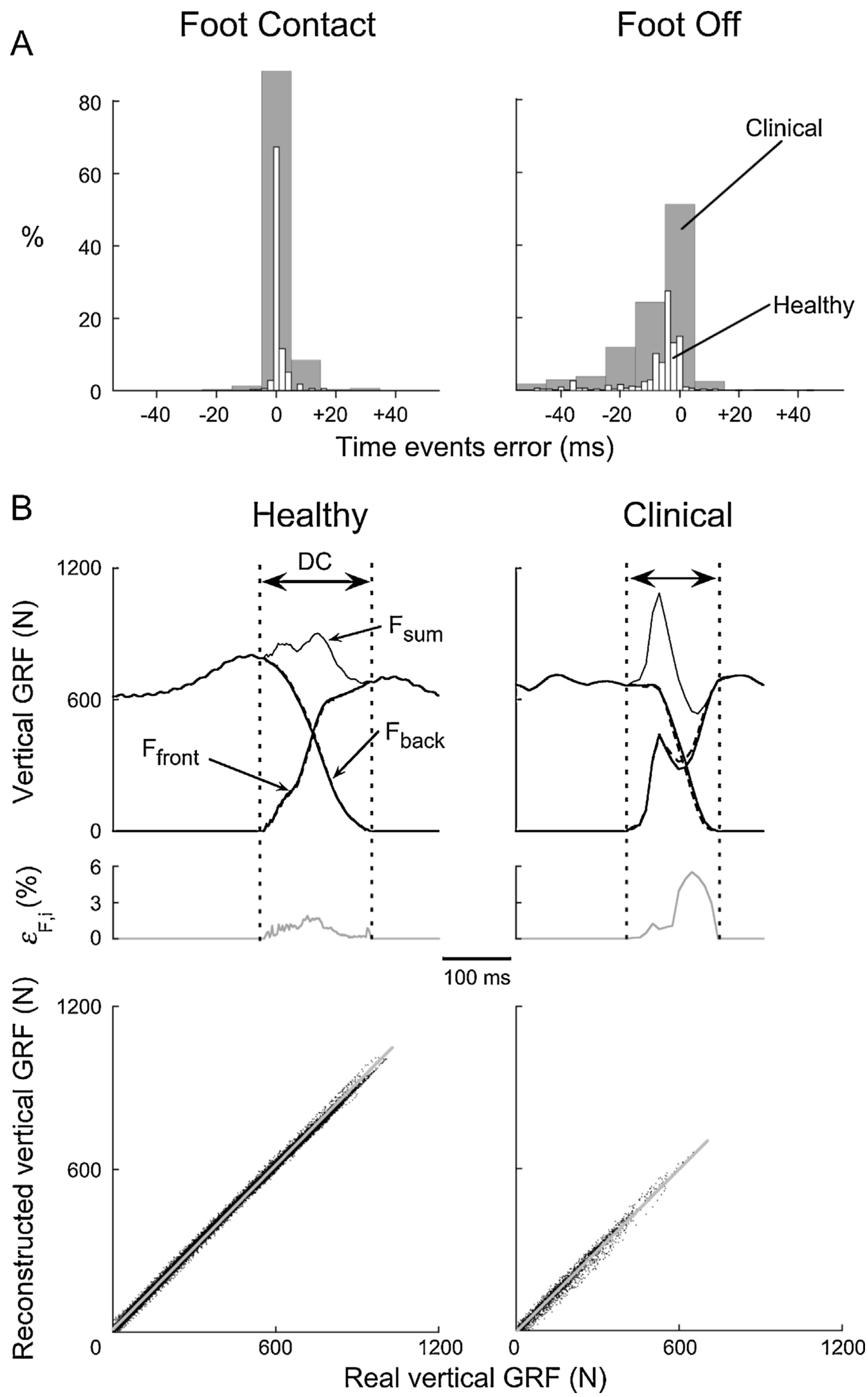

Fig. 3. DC detection and GRF reconstruction.
(A) Time events error distribution of the DC detection with the current method. Time event errors are expressed in milliseconds. The white and grey bars represent the healthy (n = 374) and the clinical data (n = 437), respectively.
(B) Ground reaction forces, relative error and correlation between the real and reconstructed GRFs during the double contact (Top) Vertical ground reaction forces (GRFs) as a function of time. The bold continuous lines represent the real GRFs acting under the front and back limbs (respectively F_{front} and F_{back}), whereas the dashed lines represent the reconstructed GRFs. Note that the dashed lines are partially hidden by the bold continuous lines. The thin continuous line represents the sum of F_{front} and F_{back} (F_{sum}). (Middle) Time evolution of the relative error ($\varepsilon_{F,i}$) during DC. Same traces as in Fig. 1. (Bottom) Correlation between the real and reconstructed GRFs acting upon the back limb for healthy and clinical gait during DC. The data points represent all samples during DC. The grey lines represent the linear regressions. Healthy: n = 23073, R^2 = 0.998, slope = 1.007 [1.006; 1.007], intercept = 9.755 [9.516; 9.995], $p < 0.001$. Clinical: n = 5149, R^2 = 0.991, slope = 1.010 [1.007; 1.013], intercept = −6.320 [−7.122; −5.518], $p < 0.001$. The values in brackets specify the 95% confidence interval of the slope and intercept coefficients.

Villeger et al. [25] adapted the detection method of Verkerke et al. [9]. The authors used the forward CoP speed to detect DC with the mean walking speed as a threshold. This method is easy to implement and performs well on FC. Nevertheless, it is less efficient for FO because, on some cases, the forward CoP speed presents multiple threshold-crossings during DC.

Following the drawbacks of previous studies mentioned above, we developed a new method of detection for FC and FO based on the minimum and maximum of d_{path} (Fig. 1). Within a single step cycle, the minimum and maximum are, by definition, always present and thus allow an automatic and systematic detection of DC limits. We emphasize that our methodology suits any walking pattern, does not rely on subjective appreciation and does not need the data of a previous step. Also, our methodology does not require a threshold to detect FC and FO and thus limits the risk of false detections or, on the contrary, the complete lack of detection.

Our results suggest a reliable detection of DC with 75% of our data presenting no error for FC detection and only 10 ms or less error on FO (Table 2, P_{75}). Note that the values reported are dependent on the time interval between two samples. In our case, the sample rate for the clinical gait trials was 100 Hz, which means an error resolution as large as 10 ms; better results can be expected with higher sample rates. Moreover, our absolute errors are the smallest reported in the literature. As mentioned previously, the FC and FO detections are of first importance but are only the prime stage to the force reconstruction.

In order to estimate the individual vertical forces based on kinetic data, Ballaz et al. [23] used a spline interpolation between FC and FO. Villeger et al. [25] used a more complex mathematical model taking into account the walking speed and pre-DC GRF characteristics. Anyhow, for vertical forces, both studies report a similar reconstruction error of 3.8%. Clearly, the 'interpolation' methods allow a valuable force reconstruction for healthy gait because it does reproduce the smooth weight-transfer from the back to the front limb that occurs during DC. However, an interpolation is less suitable for clinical gait (Table 2) where the transition is not always smooth and thus, no longer adjustable to a spline or other curve fitting mathematical model.

While applying the Davis and Cavanagh's methodology to our set of data, the need for improvements and clarifications arose. First, our equations are written with force parameters that are commonly accessible: CoP and total vertical force measured by the device. Second, the moments are calculated about an axis that maximizes the lever arms and thus improves the force reconstruction quality and, most important, allows a force reconstruction in all cases. Actually, Davis and Cavanagh computed the moments about the fore–aft axis and this can lead to erroneous or even failure to reconstruct the individual forces. In addition, these authors tested their reconstruction method on only one subject and since then, to the best of our knowledge, nobody has reported validation of the individual force reconstruction using the same methodology.

Our results highlight a method of reconstruction more effective than those proposed in the literature; for instance, the reconstructed and true forces are highly correlated ($R^2 > 0.99$, Fig. 3B), the relative error presents a positive skewed distribution and is below 3% for three quarters of our data (Table 2). It is also important to stress out that the error is only due to the fact that CoP_{back} and CoP_{front} are estimated from pre- and post-DC CoP characteristics with constant values during the whole period of DC, which is not the case in reality. Indeed, the error is nulled when the method is used with the real CoP_{back} and CoP_{front} measured during DC instead of the fixed values. A CoP interpolation under each foot would decrease the error but may also not be applicable to clinical gait. If any, this interpolation should be implemented with caution.

In summary, this study presents an automatic and accurate method for the detection of FC and FO in order to reconstruct the vertical GRFs acting upon each limb during DC. This algorithm has been validated on 811 steps, at different speeds ranging from 0.83 to 1.94 m s^{-1} and on healthy and clinical gaits. Further studies may use this decomposition algorithm with confidence when the evaluation of the force acting upon each lower limb is required, for instance using a single force plate, an instrumented treadmill, or when the foot lands on more than one force plate.

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Conflict of interest

The authors have no conflict of interest to declare.

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